OPTICAL SPECTROSCOPY PATHLENGTH MEASUREMENT SYSTEM

Reference to Prior Applications

[0001] This Application is a continuation of Application 09/925,982, filed August 9, 2001 and is related to and claims the benefit of priority under 35 U.S.C. § 119(e) from U.S. Provisional Application No. 60/226,428 filed on August 18, 2000.

Background of the Invention

[0002] Various optical spectroscopic measurement systems have been developed for the noninvasive monitoring of blood constituent concentrations. In such systems, light of multiple wavelengths is used to illuminate a thin tissue portion of a person, such as a fingertip or earlobe, to obtain a spectrum analysis of the light absorbed by blood flowing through the tissue site. Pulse oximetry systems, which perform such measurements to monitor blood oxygenation of hemoglobin constituents, have been particularly successful in becoming the standard of care. Extending this technology to the noninvasive monitoring of other blood constituents, such as blood glucose, is highly desirable. For example, current methods for accurately measuring blood glucose involve drawing blood from the subject, which can be onerous for diabetics who must take frequent samples to closely monitor blood glucose levels.

[0003] FIG. **1** illustrates an optical spectroscopic measurement system **100**. A multiple wavelength light source **110** produces incident light **112** of intensity I_0 and wavelength λ, $I_{0,\lambda}$, which illuminates a sample **120** having multiple constituents, each of concentration c_i. The incident light **112** is partially absorbed by the sample **120**, and transmitted light **130** of intensity I emerges from the sample **120**. A detector **140** provides an output signal **142** that is proportional to the transmitted light **130**. A signal processor **150** operates on the detector output signal **142** to provide a measurement **152** that is indicative of one or more of the constituent concentrations c_i in the sample **120**, based upon the known extinction coefficients $\varepsilon_{i,\lambda}$ of the sample constituents.

Summary of the Invention

[0004] The attenuation of light through a homogenous, non-scattering medium of thickness d having n dissolved, absorbing constituents is described by the Beer-Lambert Law

$$I = I_{0,\lambda} e^{-\sum_{i=1}^{n} \varepsilon_{i,\lambda} \cdot c_i \cdot d} \tag{1}$$

Dividing both sides by $I_{0,\lambda}$ and taking the logarithm yields

$$\ln(I/I_0) = -\mu_a \cdot d \tag{2a}$$

$$\mu_a = \sum_{i=1}^{n} \varepsilon_{i,\lambda} \cdot c_i \tag{2b}$$

where μ_a is the bulk absorption coefficient and represents the probability of absorption per unit length. Measurements are taken at n wavelengths to yield n equations in n unknowns

$$\begin{bmatrix} \ln(I_{\lambda_1}/I_{0,\lambda_1}) \\ \ln(I_{\lambda_2}/I_{0,\lambda_2}) \\ \vdots \\ \ln(I_{\lambda_n}/I_{0,\lambda_n}) \end{bmatrix} = -\begin{bmatrix} \varepsilon_{1,\lambda_1} & \varepsilon_{2,\lambda_1} & \cdots & \varepsilon_{n,\lambda_1} \\ \varepsilon_{1,\lambda_2} & \varepsilon_{2,\lambda_2} & \cdots & \varepsilon_{n,\lambda_2} \\ \vdots & \vdots & \ddots & \vdots \\ \varepsilon_{1,\lambda_n} & \varepsilon_{2,\lambda_n} & \cdots & \varepsilon_{n,\lambda_n} \end{bmatrix} \begin{bmatrix} c_1 \\ c_2 \\ \vdots \\ c_n \end{bmatrix} d \tag{3}$$

which can be written in matrix notation as

$$\mathbf{I} = -\mathbf{A}(\lambda)\mathbf{C}d \tag{4}$$

Solving for the constituent concentrations yields

$$\mathbf{C} = -\frac{1}{d}\mathbf{A}(\lambda)^{-1}\mathbf{I} \tag{5}$$

[0005] If the medium is a tissue portion of a person, such as a fingertip, it includes a number of constituents that absorb light. Some of the principal absorbing constituents in tissue include water, oxyhemoglobin, reduced hemoglobin, lipids, melanin and bilirubin. A drawback to applying the Beer-Lambert Law to determine the concentrations of absorbing constituents, however, is that tissue is a turbid media, i.e. strongly scatters light, which violates an underlying assumption of equation (1). Scattering in

tissue is due, in part, to the variations in refractive index at the boundaries of cells or other enclosed particles, such as collagen fibers, mitochondria, ribosomes, fat globules, glycogen and secretory globules.

[0006] FIG. **2** illustrates a particular photon path **200** as it travels through a turbid medium **202**. The photon path **200** is shown as a series of connected vectors $\vec{pl}_i$ each representing the direction and pathlength of a particular photon between collisions. The total pathlength traveled by the photon is

$$pl = \sum_{i=1}^{n} \sqrt{pl_{ix}^2 + pl_{iy}^2 + pl_{iz}^2} \qquad (6)$$

[0007] As shown in FIG. **2**, the effect of scattering is to substantially increase the photon pathlength and, hence, the probability of absorption. Thus, when a turbid media is considered, the Beer-Lambert Law is modified to include the effective pathlength, pl, which is a function of wavelength. The Beer-Lambert Law is also written in terms of transmission, T, to differentiate reflected light due to back-scattering of the incident light.

$$T = T_{max} e^{-\sum_{i=1}^{n} \varepsilon_{i,\lambda} \cdot c_i \cdot pl_\lambda} \qquad (7)$$

where T_{max} is the maximum transmitted light without absorption.

$$\begin{bmatrix} \ln\left(T_{\lambda_1} / T_{max,\lambda_1}\right) \\ \ln\left(T_{\lambda_2} / T_{max,\lambda_2}\right) \\ \vdots \\ \ln\left(T_{\lambda_n} / T_{max,\lambda_n}\right) \end{bmatrix} = -\begin{bmatrix} pl_{\lambda_1} & 0 & 0 & 0 \\ 0 & pl_{\lambda_2} & 0 & 0 \\ 0 & 0 & \ddots & 0 \\ 0 & 0 & 0 & pl_{\lambda_n} \end{bmatrix} \begin{bmatrix} \varepsilon_{1,\lambda_1} & \varepsilon_{2,\lambda_1} & \cdots & \varepsilon_{n,\lambda_1} \\ \varepsilon_{1,\lambda_2} & \varepsilon_{2,\lambda_2} & \cdots & \varepsilon_{n,\lambda_2} \\ \vdots & \vdots & \ddots & \vdots \\ \varepsilon_{1,\lambda_n} & \varepsilon_{2,\lambda_n} & \cdots & \varepsilon_{n,\lambda_n} \end{bmatrix} \begin{bmatrix} c_1 \\ c_2 \\ \vdots \\ c_n \end{bmatrix} \qquad (8)$$

$$\mathbf{T} = -\mathbf{X}(\lambda)\mathbf{A}(\lambda)\mathbf{C} \qquad (9)$$

$$\mathbf{C} = -\mathbf{A}(\lambda)^{-1}\mathbf{X}(\lambda)^{-1}\mathbf{T} \qquad (10)$$

[0008] FIG. **3** illustrates one method of measuring the effective pathlength through a sample. A picosecond pulse laser **310** and an ultra-fast detector **340** directly measure the photon "time of flight" through a sample **320**. A single pulse **360** with a duration on the order of a picosecond is directed through the sample **320**. The detector **340**

measures the corresponding impulse response **370**. The time difference between the light entering the sample **312** and the mean time of flight, $\bar{t}$ **380**, of light having traversed the sample **330** yields the mean optical pathlength, i.e. the effective pathlength

$$mpl = {}^{c_v \bar{t}}\!\Big/_{n_s} \tag{11a}$$

$$\bar{t} = \int_0^\infty T(t)t\,dt \Big/ \int_0^\infty T(t)\,dt \tag{11b}$$

where c_v is the speed of light in a vacuum and n_s is the refractive index of the sample. An analytic expression for the shape of the impulse response of a narrow collimated pulsed light beam normally incident on the surface of a semi-infinite homogeneous tissue slab of thickness d, derived from the diffusion approximation to radiative transfer theory, is

$$T(d,t) = (4\pi Dc)^{-1/2} t^{-3/2} e^{-\mu_a ct} f(t) \tag{12a}$$

$$f(t) = \Big\{(d - z_0)e^{-\left[\frac{(d-z_0)^2}{4Dct}\right]} - (d + z_0)e^{-\left[\frac{(d+z_0)^2}{4Dct}\right]}$$

$$+ (3d - z_0)e^{-\left[\frac{(3d-z_0)^2}{4Dct}\right]} - (3d + z_0)e^{-\left[\frac{(3d+z_0)^2}{4Dct}\right]}\Big\} \tag{12b}$$

$$D = \{3[\mu_a + (1-g)\mu_s]\}^{-1} \tag{12c}$$

$$z_0 = [(1-g)\mu_s]^{-1} \tag{12d}$$

where *T(d, t)* is the spatially integrated transmittance, D is the diffusion coefficient, c is the speed of light in the tissue, μ_a is the bulk absorption coefficient, μ_s is the bulk scattering coefficient and g is the anisotropy, which is the mean cosine of the scattering angle. Equations (12a)-(12d), therefore, are an approximation of the impulse response **370** shown in FIG. **3**. The derivation of equations (12a)-(12d) and a description of the model upon which

that derivation is based, is given in *Time Resolved Reflectance and Transmittance for the Noninvasive Measurement of Tissue Optical Properties*, Patterson et al., Applied Optics, Vol. 28, No. 12, June 15, 1989, Optical Society of America, incorporated in its entirety by reference herein. The Patterson article also provides an expression for the mean pathlength

$$mpl = (4\mu_a D)^{-1/2} \frac{(d - z_0)e^{(2z_0\sqrt{\mu_a/D})} - (d + z_0)}{e^{(2z_0\sqrt{\mu_a/D})} - 1} \qquad (13)$$

[0009] As equation (13) indicates, the mean pathlength is dependent on geometry and the concentration of various blood constituents and dynamically changes in tissue as the geometry and blood concentration changes. A way of dynamically determining the mean pathlength through a tissue sample is needed in order to reasonably estimate constituents such as blood glucose. Unfortunately, a measurement system such as described with respect to FIG. **3**, above, is both large and expensive, confining its use to optical laboratories rather than clinical use. Instead, the pathlength measurement system of the present invention estimates the mean pathlength by measuring the magnetically-induced optical rotation of polarized light as it passes through a tissue sample.

[0010] One aspect of the present invention is a physiological monitor for measuring a blood constituent concentration within a tissue portion of a subject. The monitor has a polarized light source adapted to illuminate the tissue portion with an incident light beam and a magnetic field generator configured to impose a magnetic field on the tissue portion while illuminated by the light source. The magnetic field imparts a rotation in the plane of polarization of the incident light beam as it propagates through the tissue portion and emerges as a transmitted light beam. The monitor also has a polarimeter with an input responsive to the transmitted light beam and an output corresponding to the rotation. The monitor further has a signal processor in communications with the polarimeter output so as to compute an output corresponding to a mean pathlength estimate of the tissue portion. In one embodiment of the physiological monitor, the polarized light source and the polarimeter are adapted to provide spectroscopic measurements of the tissue portion, and the signal processor combines those spectroscopic measurements with corresponding mean pathlength estimates to provide an output indicative of the blood constituent concentration. In another

embodiment of the physiological monitor, a separate spectrometer provides the spectroscopic measurements of the tissue portion and the signal processor provides corresponding mean pathlength estimates that are combined with the spectroscopic measurements to indicate the blood constituent concentration.

[0011] In yet another embodiment of the physiological monitor described in the above paragraph, the magnetic field generator alternately imposes a plurality of magnetic fields on the tissue portion. A first one of the fields encodes with a first rotation those photons traveling through the sample generally on-axis with the light beam. A second one of the fields encodes with a second rotation those photons traveling through the sample generally off-axis with the light beam. The mean pathlength measurement is a function of the second rotation relative to the first rotation. In a particular embodiment, the first one of the fields is a uniform field coaxial with the incident light beam and the second one of the fields is a uniform field orthogonal to the first one of the fields. In another particular embodiment, the first one of the fields is a uniform field coaxial with the incident light beam and the second one of said fields is a non-uniform field coaxial with the incident light beam. In yet another particular embodiment, the mean pathlength measurement is a ratio of the second rotation to the first rotation. In a further embodiment of the physiological monitor, the magnetic field generator alternately imposes a plurality of orthogonal magnetic fields on the tissue portion and the mean pathlength estimate is a function of a corresponding plurality of rotations in the plane of polarization of the incident light beam imparted by the fields. In a particular embodiment, the function is proportional to a square-root of a sum of the squares of the rotations.

[0012] Another aspect of the present invention is a physiological monitor for measuring a blood constituent concentration within a tissue portion of a subject having a light source adapted to illuminate the tissue portion with a monochromatic light polarized in a first direction and a magnetic field generator configured to alternately impose a first magnetic field and a second magnetic field on the tissue portion while illuminated by the light. The first field imparts a first rotation on the light and the second field imparts a second rotation on the light. The monitor also has a detector responsive to light intensity polarized in a second direction. The detector provides a first output corresponding to the first rotation and a second

output corresponding to the second rotation so as to compensate for scattering in the tissue portion when calculating a blood constituent concentration. In one embodiment, the magnetic field generator is a Helmholtz coil configured to generate a first uniform magnetic field coaxially to the light source and a second uniform magnetic field orthogonally to the first uniform magnetic field. In another embodiment, the magnetic field generator is a pair of generally planar permanent magnets. The magnets are fixedly mounted parallel to each other and are rotatable between a first position that generates a first uniform magnetic field coaxially to the light source and a second position that generates a second uniform magnetic field orthogonally to the first uniform magnetic field. In yet another embodiment, the magnetic field generator is a pair of generally planar permanent magnets. The magnets are each hinged to move between a first position parallel to each other so as to generate a first uniform magnetic field coaxially to the light source and a second position tilted towards each other so as to generate a second non-uniform magnetic field coaxial to the light source.

[0013] A further aspect of the present invention is a physiological monitoring method for measuring a blood constituent concentration within a tissue portion of a subject. The method comprises the steps of illuminating the tissue portion with a polarized light beam, applying a magnetic field to the tissue portion, measuring a rotation in polarization of the light beam after transmission through said tissue portion, estimating a mean photon pathlength from the rotation and applying the mean pathlength to a spectroscopic measurement to determine the constituent concentration. In one embodiment, the method also comprises the steps of measuring an attenuation of light transmitted through the tissue portion and estimating an absorption from the result of the measuring. The applying step has the substeps of combining the mean photon pathlength and the absorption to compute a constituent concentration. In another embodiment, the method also has the steps of applying a second magnetic field to the tissue portion, measuring a second rotation in polarization of the light beam after transmission through the tissue portion, and calculating a ratio of the rotation and the second rotation so as to estimate a mean path length.

[0014] Yet another aspect of the present invention is a physiological monitor for measuring a blood constituent concentration within a tissue portion of a subject. The monitor comprises a light source means for illuminating the tissue portion with a polarized

light beam and a generator means for imparting a rotation of the polarized light beam as it propagates through the tissue portion. The monitor also comprises a detector means for outputting a measure of the rotation and a processor means for utilizing the measure to provide a compensation for scattering within the tissue portion. In one embodiment, the monitor further comprises a spectroscopic measurement means for providing an estimate of a blood constituent concentration within the tissue portion and a compensation means for combining the compensation with the estimate to improve the estimate.

Brief Description of the Drawings

[0015] FIG. **1** is an illustration depicting a conventional optical spectroscopic measurement system for determining the concentration of various constituents within a sample;

[0016] FIG. **2** is an illustration depicting a photon path through a turbid medium and a component of the photon path projected onto orthogonal components of a uniform B-field;

[0017] FIG. **3** is a diagram illustrating the mean optical pathlength through a sample;

[0018] FIG. **4** is a block diagram of an optical spectroscopy pathlength measurement system according to the present invention;

[0019] FIGS. **5A-D** are graphical representations of a photon path within a uniform coaxial B-field, within uniform orthogonal B-fields and within a non-uniform coaxial B-field, respectively;

[0020] FIGS. **6A-B** are illustrations depicting particular embodiments of an optical spectroscopy pathlength measurement system incorporating a faraday rotation modulator and a photoelastic modulator, respectively;

[0021] FIG. **7** is a perspective view of a polarimeter portion of an optical spectroscopy pathlength measurement system;

[0022] FIG. **8** is a perspective view of one embodiment of an optical spectroscopy pathlength measurement system utilizing a triaxial Helmholtz coil for magnetic field generation;

[0023] FIG. **9** is a perspective view of another embodiment of an optical spectroscopy pathlength measurement system utilizing a permanent magnet pair for magnetic field generation;

[0024] FIGS. **10A-B** are perspective views of another embodiment of an optical spectroscopy pathlength measurement system utilizing a permanent magnet pair that has a parallel position (FIG. **10A**) and an angled position (FIG. **10B**);

[0025] FIG. **11** is a perspective view of a blood constituent measurement instrument incorporating an optical spectroscopy pathlength measurement system according to the present invention; and

[0026] FIG. **12** is a functional block diagram of a blood constituent measurement instrument.

Detailed Description of the Preferred Embodiment

[0027] FIG. **4** is a block diagram of an optical spectroscopy pathlength measurement system **400** according to the present invention. The pathlength measurement system **400** has a polarized light source **410**, a magnetic field generator **420**, a polarimeter **430** and a signal processor **440**. The polarized light source **410** is adapted to illuminate a sample **402**, which may be a tissue portion of a person, such as a fingertip. The polarimeter **430** is adapted to receive transmitted light **404** through the sample **402** and measure the polarization components of that light. The polarimeter output **432** is a measure of the optical rotation produced by the sample **402**. The signal processor **440** inputs the polarimeter output **432** and provides an output **442** which is a computed estimate of the mean pathlength. Although this embodiment of a pathlength measurement system is based upon measuring light transmitted through the sample **402**, as described by equation (7), one of ordinary skill in the art will recognize that a pathlength measurement system adapted to measure reflected light from the sample **402** could also be utilized for estimating mean pathlength and for measuring blood constituents.

[0028] As shown in FIG. **4**, the magnetic field generator **420** is configured to generate a B-field **422** within the sample **402** during illumination by the polarized light source **410**. The B-field **422** is imposed on the sample **402** to create a "Faraday effect," which

causes rotation in the plane of polarization of a linearly-polarized beam of light as it propagates through a medium in the presence of a magnetic field. Due to the Faraday effect, the B-field **422** rotates the plane of polarization of the light source **410**. The amount of optical rotation is given by

$$\phi = \int V\vec{B} \bullet d\vec{l} \qquad (14)$$

where V is the Verdet constant, which depends on the material and the wavelength. The direction of rotation depends on whether light is traveling parallel or anti-parallel to the B-field. Hence, the rotation is cumulative, i.e. does not reverse when the direction of propagation reverses. One or more B-fields **422** can be used to differentially encode with rotation those photons traveling generally on-axis and generally off-axis through the sample **402**. Optical rotation is measured by the polarimeter **430** and used by the signal processor **440** to estimate the mean optical pathlength, *mpl*, of the sample **402**, as described in further detail below.

[0029] Shown in FIG. **4**, the polarized light source **410** and polarimeter **430** can be configured as a variable wavelength light source **110** (FIG. **1**) and a detector **140** (FIG. **1**) for making simultaneous spectroscopic measurements. Alternatively, the pathlength measurement system **400** can be integrated with a separate spectroscopic measurement system **100** (FIG. **1**). In either case, the pathlength estimates provided by the pathlength measurement system **400** can be combined with spectroscopic measurements to compensate for scattering in the sample **402** when estimating absorbing constituent concentrations in the sample **402**.

[0030] Also shown in FIG. **4**, without scattering, the transmitted light **404** emerging from the sample **402** has the same polarization as the incident light **412** emitted from the polarized light source **410**. The cumulative effect of scattering in the sample **402**, however, is to depolarize a significant portion of the incident light **412**. The amount of depolarization is a function of the amount of scattering that occurs. To account for this depolarization, the polarimeter output **432** can be normalized by the ratio of the intensities of the polarized light to the unpolarized light or the ratio of the polarized light intensity to the total intensity.

[0031] As FIG. **2** illustrates, when a uniform magnetic field $\vec{B}$ **210** is applied to a medium **202** an optical rotation ϕ_i is imparted to the photon while traversing a particular path segment $\vec{pl}_j$ **220**. The optical rotation is computed from the projection of $\vec{pl}_j$ onto $\vec{B}$ according to equation (14), which can be written in terms of the orthogonal components of $\vec{pl}_j$ and $\vec{B}$ as

$$\phi_j = V\sqrt{(B_x pl_{jx})^2 + (B_y pl_{jy})^2 + (B_z pl_{jz})^2} \tag{15}$$

The total optical rotation through the sample is

$$\phi = \sum_{i=1}^{n} \phi_i \tag{16}$$

[0032] FIGS. **5A-5C** illustrate the alternate application of uniform orthogonal B-fields, $\vec{B}_x$ **510** (FIG. **5A**), $\vec{B}_y$ **520** (FIG. **5B**), and $\vec{B}_z$ **530** (FIG. **5C**) to a photon path **200** through a sample **202** (FIG. **2**). This results in three optical rotation measurements, which according to equation (15) and equation (16) are

$$\phi_x = VB_x \sum_{i=1}^{n} |pl_{ix}|; \qquad \phi_y = VB_y \sum_{i=1}^{n} |pl_{iy}|; \qquad \phi_z = VB_z \sum_{i=1}^{n} |pl_{iz}| \tag{17}$$

The pathlength as expressed in equation (6) can be estimated as follows

$$pl \approx \sqrt{\left(\sum_{i=1}^{n} |pl_{ix}|\right)^2 + \left(\sum_{i=1}^{n} |pl_{iy}|\right)^2 + \left(\sum_{i=1}^{n} |pl_{iz}|\right)^2} \tag{18}$$

which underestimates the sum of the magnitudes of the individual pathlength vectors by the magnitude of the sum of the vector projections along each orthogonal axis. Combining equations (17) and (18) yields

$$mpl \approx \frac{1}{VB}\sqrt{\phi_x^2 + \phi_y^2 + \phi_z^2} \tag{19}$$

which provides an estimate of the mean path, *mpl*, in terms of the measured optical rotations due to alternately applied orthogonal, uniform B-fields of equal magnitude *B*. Assuming that the incident beam is along the x-axis and that scattering in the sample is uniform off axis,

equation (19) can be simplified with an estimate of mean pathlength based on the alternate application of a uniform on-axis B-field and an orthogonal off-axis B-field

$$mpl \approx \frac{1}{VB}\sqrt{\phi_x^2 + 2\phi_y^2} \tag{20}$$

Assuming that the incident beam is along the x-axis and the dominate term is the projection of the photon paths onto the x-axis, equation (20) can be further simplified with an estimate of mean path length based on the application of a uniform on-axis B-field

$$mpl \approx \frac{\phi_x}{VB} \tag{21}$$

[0033] A problem with the mean pathlength estimates expressed in equations (19) through (21) is the dependence on the Verdet constant, which varies with the sample constituents and wavelength. Accordingly, an alternative mean pathlength estimate can be expressed as a ratio that is used as a multiplier of the geometric pathlength d to account for the increased optical pathlength due to scattering

$$mpl \approx d\alpha\rho \tag{22}$$

where α is a constant that cancels when a ratio of constituent concentrations is computed, as described with respect to FIG. **12**, below. In one embodiment, the ratio ρ is

$$\rho = \frac{\sqrt{\phi_x^2 + \phi_y^2 + \phi_z^2}}{\phi_x} \tag{23}$$

so that the estimate of the mean path length is increased as the off-axis rotation components become more significant. An alternative estimate, which is simpler to measure is

$$\rho = \frac{\phi_y}{\phi_x} \tag{24}$$

[0034] FIG. **6A** is an illustration depicting a Faraday rotator modulator embodiment of an optical spectroscopy pathlength measurement system **600** according to the present invention. The pathlength measurement system **600** has a polarized light source **610** and an optical rotation sensor **620**. The polarized light source **610** and the optical rotation sensor **620** together form a polarimeter. The polarized light source **610** illuminates a sample **630**, which may be a tissue portion of a person, such as the fingertip shown. The optical rotation sensor **620** receives light transmitted through the sample **630**. A magnet field

generator **640** creates a B-field **642** within the sample **630** during illumination by the light source **610**.

[0035] Also shown in FIG. **6A**, the polarized light source **610** has a multiple wavelength optical emitter **612**, a first polarizer **614**, and a modulator **616**. The optical rotation sensor **620** has a second polarizer **624** and a detector **628**. The emitter **612** radiates a beam **602** through a first polarizer **614**, which has its polarization axis oriented at a reference angle, shown as 0° from the horizontal plane. The electric field from the beam **602** is linearly polarized by the first polarizer **614**. In order to reduce system noise, the polarized beam **604** passes through a Faraday rotator modulator **616**. The modulator is driven by a sinusoidal source **618** that phase modulates the electric field of the polarized beam **604**. The modulated beam **605** passes through the sample **630** where it obtains a optical rotation ϕ. The rotated beam **607** then passes through the second polarizer **624**, which has its polarization axis aligned orthogonally from that of first polarizer **614**. Due to the optical rotation, the rotated beam **607** has an electrical field component along the axis of the second polarizer **624**. The output beam **609** from the second polarizer **624** is then measured by a detector **628**, such as a photodiode. The electric field of the output beam **609** can be expressed as

$$E(t) = E_0 \sin[\phi + \theta_m \sin(\omega_m t)] \tag{25}$$

where ω_m is the modulation frequency and θ_m is the modulation amplitude. Therefore, the irradiance measured by the detector **628**, which is proportional to the square of the electrical field $E(t)$ of the output beam **609** is

$$I(t) = E_0^2 (\frac{c\varepsilon_0}{2}) \sin^2[\phi + \theta_m \sin(\omega_m t)] \tag{26}$$

Equation (26) can be simplified using the following identity

$$\sin^2 x = \frac{1-\cos(2x)}{2} \tag{27}$$

From equation (26) and equation (27)

$$I(t) = E_0^2 (\frac{c\varepsilon_0}{2}) \left\{ \frac{1-\cos[2\phi + 2\theta_m \sin(\omega_m t)]}{2} \right\} \tag{28}$$

Assuming the angle of the cosine in equation (28) is small, the cosine may be approximated with the first two terms of its Taylor series expansion

$$\cos(x) \cong 1 - \frac{x^2}{2} \tag{29}$$

As a result, the detector output **652** is

$$I(t) = \kappa_d E_0^2 (\frac{c\varepsilon_0}{2}) \left\{ \phi^2 + 2\phi\theta_m \sin(\omega_m t) + \frac{\theta_m^2 [1 - \cos(2\omega_m t)]}{2} \right\} \tag{30}$$

where κ_d is the detector gain. The desired quantity, ϕ, occurs at the modulation frequency, which can be obtained by high pass filtering the DC term and using a lock-in amplifier (not shown) from the detector output **652** to control the sinusoidal source **618**, as is well-known in the art. Thus, the detector output **652** provides a signal proportional to the optical rotation within the sample **630**.

[0036] FIG. **6B** is an illustration depicting a photoelastic modulator (PEM) embodiment of an optical spectroscopy pathlength measurement system **660** according to the present invention. The pathlength measurement system **660** has a polarized light source **610**, and an optical rotation sensor **620**, and a magnet field generator **640** that function generally as described above with respect to FIG. **6A**. In the embodiment of FIG. **6B**, the polarized light source **610** has a multiple wavelength optical emitter **612** and a first polarizer **614** oriented at 0°. The optical rotation sensor **620** has a PEM **622**, a second polarizer **624** oriented at 45° and a detector **628**. The emitter **612**, first polarizer **614** and detector **628** also function generally as described above with respect to FIG. **6A**. The first polarizer **614** is oriented parallel to the PEM **622** axis. If no rotation occurs, no component of the polarization at 45° to the PEM **622** axis will occur. Hence, no modulated signal will be measured by the detector **628**. Any rotation would result in a detected signal, which for small angles is proportional to the angle of rotation.

[0037] FIG. **7** illustrates the polarimeter portion of the optical spectroscopy pathlength measurement system. The polarimeter **700** has a body **710**, a light source **610** and a rotation sensor **620**. In one embodiment, the body **710** is a hollow cylinder adapted to accept a fingertip **630**. The body **710** integrates the light source **610** and the rotation sensor **620**, which have ends terminating at opposite sides of the body **710**. The light source **610** and the rotation sensor **620** are configured so that a beam from the light source **610** will

illuminate a fingertip placed within the body **710** and so that a corresponding beam emerging from the fingertip will be received by the rotation sensor **620**. Wiring **720** extending from the light source **610** and rotation sensor **620** provide power and signal paths, respectively.

[0038] FIG. **8** illustrates an embodiment of the optical spectroscopy pathlength measurement system **800** utilizing a triaxial Helmholtz coil for magnetic field generation. The pathlength measurement system **800** has a polarimeter portion **700** enclosed by a triaxial Helmholtz coil **802**. A Helmholtz coil is a pair of coaxially mounted electromagnetic coils separated by a distance equal to the coils' diameter. When excited by coil currents in the same direction, the magnetic fields add in a manner that produces a very uniform magnetic field along the common axis of the coils, as is well-known in the art.

[0039] Shown in FIG. **8** are three Helmholtz coil pairs, including an x-axis pair **812**, a y-axis pair **822**, and a z-axis pair **832** mounted along mutually orthogonal axis to form a cube. Electrical current to energize the coils is routed independently to each of the three coil pairs **812**, **822**, **832** via a cable **840**. The polarimeter **700** is oriented within the Helmholtz coil **802** so that the light beam from the light source **610** to the rotation sensor **620** is aligned with the x-axis **850** and the body **710** is coaxial with the z-axis **860**. With this configuration, each of the three coil pairs **812**, **822**, **832** can generate uniform, orthogonal B-fields within a fingertip inserted into the body **710**. Hence, the three coil pairs **812**, **822**, **832** are alternately energized during operation of the polarimeter **700** to measure the resulting optical rotations and to estimate the mean pathlength through a fingertip sample according to equation (19). With this same configuration, the x-axis coil pair **812** and the y-axis coil pair **822** are alternately energized during operation of the polarimeter **700** to measure the resulting optical rotations and to estimate the mean pathlength through a fingertip sample according to equation (20). Similarly, the x-axis coil pair **812** is energized during operation of the polarimeter **700** to measure the resulting optical rotation and to estimate the mean pathlength through a fingertip sample according to equation (21).

[0040] FIG. **9** illustrates another embodiment of an optical spectroscopy pathlength measurement system **900** utilizing a permanent magnet pair for magnetic field generation. The pathlength measurement system **900** has a polarimeter portion **700** mounted within a pair of permanent magnets **910**, each having a planar face **912**. The permanent

magnet pair **910** is connected by a crossbar **930** of nonmagnetic material, and a knob **940** is mounted atop the crossbar **930**. The polarimeter **700** is oriented between the magnets **910** so that the light beam from the light source **610** to the rotation sensor **620** is aligned with the x-axis **950** and the cavity of the body **710** is coaxial with the knob **940** and the z-axis **960**. The magnets **910** are rotatable about the z-axis **960** from a first position where the faces **912** are normal with the x-axis **950** to a second position where the faces **912** are normal with the y-axis **970**.

[0041] With this configuration, the rotatable magnets **910** can generate uniform, orthogonal B-fields oriented along the x-axis **950** and y-axis **970** within a fingertip inserted into the body **710**. Hence, the rotatable magnets **910** are alternately rotated between the first position (shown) and the second position during operation of the polarimeter **700** to measure the resulting optical rotations and to estimate the mean pathlength through a fingertip sample according to equation (20). With this same configuration, the magnet pair **910** can remain in the first position (shown) during operation of the polarimeter **700** to measure the resulting optical rotation and to estimate the mean pathlength through a fingertip sample according to equation (21).

[0042] FIGS. **5A** and **5D** depict a photon path **200** within uniform **510** (FIG. **5A**) and non-uniform **540** (FIG. **5D**) coaxial B-fields. As shown in FIG. **5A**, a photon propagates along path **200** as the result of an incident beam aligned with the x-axis. Scattering within the illuminated medium results in the path **200** deviating off-axis in a random manner. A uniform field $\vec{B}_x$ **510** (FIG. **5A**) imparts a first rotation to the incident beam, which is measured with a polarimeter as described with respect to FIG. **6A** and FIG. **6B**, above. As shown in FIG. **5D**, a non-uniform field $\vec{B}_x{}'$ **540** imparts a second rotation to the incident beam, which is also measured with the polarimeter. The non-uniform field **540** (FIG. **5D**) is devised so that the field strength is greater off-axis. The non-uniform field **540** (FIG. **5D**) acts to encode photons that deviate more off-axis with a proportionately larger optical rotation as compared to photons that remain relatively on-axis, i.e. "snake photons." The uniform field **510** (FIG. **5A**) acts as a reference, i.e. provides a reference rotation for

unencoded photons. Thus, the following ratio can be used in conjunction with equation (22) to form an estimate of the mean path length.

$$\rho = \frac{\phi_n}{\phi_u} \qquad (31)$$

where ϕ_n and ϕ_u are the measured optical rotations in the presence of the non-uniform and the uniform B-fields, respectively.

[0043] FIGS. **10A-B** illustrate another embodiment of an optical spectroscopy pathlength measurement system **1000** utilizing a permanent magnet pair that creates both a uniform and a non-uniform magnet field. The pathlength measurement system **1000** has a polarimeter portion **700** mounted within a pair of permanent magnets **1010**. Each magnet **1010** has a generally planar face **1020** and a hinge **1030**. The hinges **1030** allow the magnets to be moved between a parallel position, as shown in FIG. **10A**, and an angled position, as shown in FIG. **10B**. The polarimeter **700** is oriented between the magnets **1010** so that the light beam from the light source **610** to the rotation sensor **620** is aligned with the x-axis **1050**, which is normal to the magnet faces **1020** in the parallel position.

[0044] With this configuration, the hinged magnets **1010** in the parallel position (FIG. **10A**) generate a uniform B-field within a fingertip inserted into the body **710**. The hinged magnets **1010** in the angled position (FIG. **10B**) generate a non-uniform B-field within the fingertip. Hence, the hinged magnets **1010** are alternately tilted between the parallel position (FIG. **10A**) and the angled position (FIG. **10B**) during operation of the polarimeter **700** to measure the resulting optical rotations and to estimate the mean pathlength through a fingertip sample according to equation (31).

[0045] FIG. **11** illustrates a noninvasive blood constituent measurement instrument **1100** incorporating an optical spectroscopy pathlength measurement system according to the present invention. The instrument **1100** has a chassis **1110** including a front panel **1120**. The front panel **1120** has a sample area **1130**, control keys **1140** and a display **1150**. The chassis **1110** houses a pathlength measurement system including a signal processor and an associated power supply. The polarimeter portion **700** (FIG. 7) is mounted to the front panel **1120** behind the sample area **1130** so that a fingertip can be inserted into the sample area **1130** for noninvasive blood constituent measurements. The associated B-

field generator **420** (FIG. **4**), such as the triaxial Helmholtz coils **802** (FIG. **8**), rotatable magnets **910** (FIG. **9**) or hinged magnets **1010** (FIGS. **10A-B**) are also mounted to the front panel **1120** proximate the polarimeter **700** (FIG. **7**). In operation, a person places their fingertip into the sample area **1130** and initiates a measurement by pressing a key **1140**. The polarimeter and magnets are activated to perform spectroscopy and mean path length measurements, as described with respect to FIG. **12**, below. The internal signal processor processes these measurements to compute a blood constituent concentration value, as described with respect to FIG. **12**, below. The blood constituent value is then provided to the person as a reading on the display **1150**.

[0046] FIG. **12** provides a functional block diagram of a noninvasive blood constituent measurement instrument **1100** (FIG. **11**). When a blood constituent measurement is initiated, the polarimeter light source **610** (FIGS. **6A-B**) illuminates the fingertip with a predetermined wavelength **1210**. Multiple wavelength optical emitter embodiments, including a broadband optical source transmitting through a filtering element are described in U.S. Patent No. 5,743,262 entitled "Blood Glucose Monitoring System," which is assigned to the assignee of the present invention and incorporated herein by reference. At the same time the fingertip is illuminated, a magnetic field is applied **1220** to induce the Faraday effect. The resulting optical rotation is measured **1230** by the rotation sensor **620** (FIGS. **6A-B**). Further, light intensity attenuation is also measured **1240** by the rotation sensor detector **628** (FIGS. **6A-B**). Depending on the method used to estimate mean pathlength, the magnetic field may be changed **1232** and further rotation measurements made **1230**. Depending on the number of constituents to be resolved, the wavelength of the light source is changed **1242** and the process described above is repeated for each wavelength used. Each of the optical rotation **1230** and attenuation measurements **1240** are filtered, amplified and digitized at the output of the rotation sensor detector **628** (FIGS. **6A-B**).

[0047] As shown in FIG. **12**, the resulting digital values from the detector are processed by a signal processor **650** (FIGS. **6A-B**). The processor applies one or more of equations (19)-(24), (31) to estimate the mean pathlength **1250** at each applied wavelength. The processor also computes the tissue sample absorption at each wavelength **1250**. The processor recalls from memory predetermined extinction properties for the tissue constituents

1260 for each applied wavelength. The processor then solves equation (10) or equivalent to determine the constituent concentrations **1270**. As noted above, these concentrations may only be determinable to within an unknown constant, which cancels when a concentration ratio **1280** is computed. For example, the ratio of the computed concentration for glucose to the computed concentration for water can be calculated to provide an absolute glucose concentration. Finally, the relative constituent concentration is provided as an output **1290** to the instrument display or to another connected medical instrument or display.

[0048] The optical spectroscopy pathlength measurement system has been disclosed in detail in connection with various embodiments of the present invention. These embodiments are disclosed by way of examples only and are not to limit the scope of the present invention, which is defined by the claims that follow. One of ordinary skill in the art will appreciate many variations and modifications within the scope of this invention.